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Efficient and damage-free ultrashort pulsed laser cutting of polymer intraocular lens implants

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ABSTRACT

Ophthalmic intraocular lenses are conventionally machined by diamond tools. A promising alternative approach is contour cutting by ultrashort pulsed laser micromachining. To improve process knowledge, a parametric study of picosecond and femtosecond laser machining of medical grade hydrophilic copolymers and PMMA is carried out. Material removal rates and machining quality with respect to main process parameters are determined. Reasons for chipping and formation of heat affected zones are identified and an optimized process strategy is derived. By choosing a defined pulse overlap, heat accumulation is kept minimal while increasing absorptivity through incubation avoids chipping.

1. Introduction to intraocular lenses

In ophthalmology, intraocular lenses (IOL) are implanted into the human eye to treat various refractive errors. The most common procedure performed by ophthalmic surgeons is cataract surgery which is the development of an opacification in the natural eye lens. In a cataract surgery an intraocular lens is implanted to replace the diseased natural lens. As cataracts occur most commonly due to aging, annually more than 20 million cataract surgeries are performed worldwide [1]. Intraocular lenses consist of two main functional parts: the centrally located refractive surface with diameter of 6–11 mm for vision correction and the outer region of the lens, called haptics with diameter of 11–15 mm. The haptics allows attachment of the lens in different eye parts which depends on the refractive error to be treated. As shown in Fig. 1, various designs of haptics have been developed to be able to react to the given physiological properties of diverse eye tissues to ensure high rotational stability and small displacement of the lens [2]. An important feature of modern IOL are sharp edges around the posterior optical surface which is in contact with epithelial cell. A sharp edge is proven to prevent migration of cells on the optical surface causing posterior capsule opacification which occurs in case of round edges with a higher probability [3].

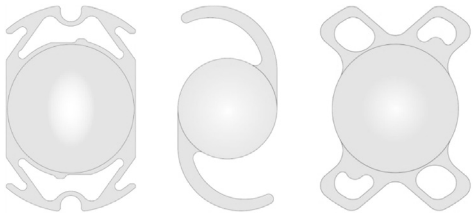

Fig. 1. Various versions of currently available intraocular lenses.

Intraocular lenses are made of conventional PMMA or modern hydrophobic or hydrophilic acrylate polymers [4]. The high brittleness (2–6% elongation at break) of such polymers and high precision required for mass-production of the IOL's optical surfaces and sophisticated haptic forms represent challenges to common manufacturing processes. Conventionally, intraocular lenses are made by machining techniques using diamond tools. First, the intraocular lens surface is created by diamond cutting followed by subsequent diamond milling of the haptics contour. This provides a polish-free optical-grade quality with sub-nm resolution of the central optic surface and the haptics edges. However, main disadvantages are that tool wear causes increasing deterioration of the machining results towards the end of the tool lifetime. Particularly, milling of the contour is slow and very time-consuming taking about 5 min per lens due to the low applicable feed rate of 30–90 mm/min which is crucial for maintaining low heat input, controlled material removal and small haptic deformation by a small applied force [5]. In addition, the minimum detail size of the generated IOL contour is limited by the diameter of the milling tool. The minimum available diameter of conventional IOL diamond milling cutter is about 0.3 mm and thus, limits the flexibility in the development of novel haptic designs.

One possible production tool that offers fast processing is the CO_2 laser which has found many applications in processing of polymers. However, the minimum achievable kerf width of several hundred micrometres [6] and large heat affected zones are crucial disadvantages of this approach. The laser beam is fully absorbed at the surface causing melting, evaporation and resolidification of material. Due to heat conduction inacceptable large heat affected zones >50 μm occur and can affect the small haptic structures

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which are partially of similar size due to reduced strength and chemical transformations.

An alternative microfabrication method is material removal by means of ultrashort pulsed laser ablation. Picosecond (ps) and femtosecond (fs) laser ablation can result in shallow heat affected zones generated around and below the processed surface and thus, is perfectly suited for processing of heat sensitive materials. As a result, this approach has become established in the production of medical stents [7], chemically strengthened glass for mobile phone displays [8] or the structuring of polycrystalline diamond for micro-drills [9]. There are also few previous studies on ultrashort pulse laser processing of polymers such as fine cutting of bio-absorbable polymers [10] and micro structuring of IOL copolymers [11]. In summary, high processing quality and unique process approaches can be realised but comprehensive experimental studies and optimization efforts are required. Although great progress has been made in research of ultrashort laser material interaction, the individual process development is still overcomplicated for instance by nonlinear optical effects and absorption in transparent materials as well as distinct multi-pulse interaction mechanisms for both metals and dielectrics [12]. Suitable laser systems for material processing are shown to be high-power ps or fs lasers. Current laser sources are designed for use in industrial environments and have great potential for the economic processing due to their high pulse energies >100 μJ available even at laser pulse repetition rates up to several 100 kHz. For such high-power laser processing quality strongly depends on the used laser system which is mainly defined by the pulse duration and likewise by the parameter set consisting of a variety of adjustable parameters. Especially, at high pulse energies combined with high pulse repetition rates and small lateral pulse to pulse separation heat accumulation as well as nonlinear optical effects occurs. This can limit the achievable processing speed, though high throughput is crucial for economic IOL mass production.

Issued by these challenges, this paper investigates the feasibility of machining IOL polymers by ultrashort laser pulses and compares ps and fs pulse duration for production of IOL haptics. The required understanding of the highly nonlinear process of micromachining by means of ultrashort laser pulses is generated by pulsed laser ablation experiments accompanied by subsequent qualitative and quantitative evaluation of the material removal results. High-speed camera monitoring is utilized to reveal process characteristics and enable visual perception of laser-material interaction.

2. Materials and experimental methods

In the laser ablation experiments, 0.35 mm thick substrates of a commercially available and widely used IOL copolymer (Contaflex, Contamac Ltd.) is used which consists of 30% PMMA, 69% PHEMA and about 1% of a UV absorber. It is a hydrophilic acrylate which is hard and brittle like PMMA in dry state and becomes soft, foldable and swells when hydrated. The UV absorber absorbs light below a wavelength of 400 nm while in the visual and near infrared spectrum the material is fully transparent [13]. To prevent hydration by air moisture, the substrates are stored in containers filled with silica gel.

2.1. Experimental setup

In the present investigations, two beam sources are used for comparison. First, a Nd:YVO_4 laser system (Fuego, Time-Bandwidth Products AG) with a wavelength of 1064 nm, pulse duration of 10 ps, an average power of 25 W and maximum pulse repetition rate of 8.2 MHz and secondly, an Yb-fiber laser system (Origami — 10 XP, OneFive GmbH) with a wavelength of 1030 nm, pulse duration of 350 fs, an average power of 3.5 W and maximum pulse repetition rate of 1 MHz. As shown in Fig. 2(a), for both laser types a scanner (Scanlab AG) is used for beam guidance, which allows a high deflection speed of the focal laser spot of up to 20,000 mm/s.

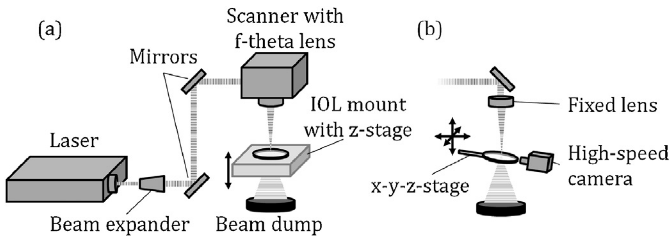

Fig. 2. (a) Setup used for carrying out the intraocular lens cutting experiments and (b) setup for high-speed camera based process monitoring.

The focal length of the f-Theta lens, which focuses the laser beam on the material surface, has a focal length of 160 mm which produces a $1/e^2$ focus diameter of $2w_0 = 31$ μm with a calculated Rayleigh length of 0.65 mm. Thus, the Rayleigh length is significantly longer than the IOL material thickness of $h = 0.35$ mm creating a nearly parallel beam path around the beam waist. This can be used for a simplified procedure as no focus tracking during ablation is necessary and enables easy sample handling and positioning while maintaining a constant intensity distribution over the whole material thickness.

In a similar setup shown in Fig. 2(b) comparable experiments are carried out with a fixed lens with 150 mm focal length instead of a scanner. Simultaneously, as the material is transparent, the process can be monitored by high-speed pump-probe shadowgraphy to provide insight into the ablation mechanisms. As shown in Ref. [14], the incidence of each pulse in the bulk of the material can be imaged with this technique by a high-speed camera (Phantom v1210 FAST, Vision Research) to acquire a qualitative estimation of the spatial distribution of energy deposition. Hereto, digitally image processed pressure waves originating from strongly heated zones are used for detection of localized energy input. To be able to identify the influences of the individual process parameters, the quantities shown in Table 1 are varied in a nearly full-factorial design of experiment. Single passes with linear feed are carried out to produce trenches followed by experiments with multiple passes creating through cuts. Main influencing parameters are pulse energy E_P, repetition rate f_P and pulse to pulse separation d_P. The feed rate is adjusted accordingly. The processed substrates are measured using a laser scanning microscope (OLS 4000, Olympus) which enables the evaluation of optical quality of the produced trenches and to measure cutting depth. Finally, complete and realistic contours with selected parameters are cut to fabricate prototypic IOL.

Table 1
Process parameter variation used in the experiments.

Laser system/pulse duration	10 ps	350 fs
Wavelength λ	1064 nm	1030 nm
Pulse energy E_P	55–112 μJ	5–35 μJ
Repetition rate f_P	1–400 kHz	1–400 kHz
Pulse separation d_P	0.1–20 μm	0.1–20 μm
Resulting feed rates $v = f_p \cdot d_p$	5–4000 mm/s	5–4000 mm/s
Resulting overlap $\Omega = 1 - d_p/2w_0$	35–99.7%	35–99.7%
Peak fluence $F_0 = 2E_p/(\pi w_0^2)$	15–30 J/cm^2	1.3–9.4 J/cm^2
Single pulse ablation threshold F_{th}	9.05 J/cm^2 [11]	3.15 J/cm^2

3. Results and discussion

3.1. Quality evaluation of ps-laser cutting

The optical evaluation of single pass ps laser ablation experiments shows that there are considerable differences in the cutting results, thus, may not be suitable for use in medical technology products and especially implants such as intraocular lenses. As shown in Fig. 3 (inlays), disqualifying characteristics are recognizable and can be classified according to their appearance which comprises clean cutting: surface chipping and heat affected zones. Thus, the results of the examined parameter range (Table 1) can be systematically displayed as shown in Fig. 3.

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